Proceedings – 23rd Annual Conference – IEEE/EMBS Oct.25-28, 2001, Istanbul, TURKEY

Hill Muscle Model Performance During Natural Activation And Electrical Stimulation

Eric J. Perreault, Thomas G. Sandercock and Charles J. Heckman
Department of Physiology, Northwestern University Medical School, Chicago, IL, USA

Abstract - **This study assessed the ability of the Hill model to predict muscle force for both electrically stimulated and naturally activated cat soleus muscle. Our results indicate that Hill model errors increase with increases in muscle velocity and decrease with increases in motor unit firing rates. For a given muscle velocity, errors were largest in for the range of firing frequencies most relevant for naturally activated muscle. During large muscle displacements, average Hill model errors often were greater than 100%.**

***Keywords* - muscle, Hill model, biomechanics**

I. INTRODUCTION

Physiologically relevant models of muscle force generation are essential for the creation of realistic large-scale simulations to examine the role of muscle properties in controlling movement and posture. Hill-like models incorporating length-tension and force-velocity properties of muscle [1, 2] have become ubiquitous in such studies. These models are attractive because of their computational simplicity and close relation to commonly measured experimental variables, but there have been surprisingly few experimental validations of Hill models during functionally relevant conditions. This study evaluated Hill model performance using functionally relevant neural inputs and muscle movements to provide bounds on the accuracy provided by such models and keys as to how these models should be improved to best simulate muscle behavior.

The most common Hill model incorporates the assumption that muscle force-velocity, length-tension, and activation properties are mutually independent, which is known to be incorrect [3]. As a consequence, such models do not capture many nonlinear muscle properties such as history-dependent effects, length and velocity dependent activation, and yielding. Therefore, numerous groups have modified the Hill model by incorporating these and other properties [4, 5]. Many of these properties, though, have been demonstrated only under laboratory conditions using length and activation inputs that typically are not seen during functional behavior. Therefore, it is not yet clear whether such complex properties are required to predict muscle force responses during normal function. One study has attempted to answer this question. Sandercock and Heckman [6] evaluated the Hill model during simulated locomotor activity in electrically stimulated cat soleus muscle. Their results indicated Hill model errors were moderate during muscle stimulation (< 10%), but increased to approximately 30% during muscle relaxation.

This study extends the approach used by Sandercock and Heckman to assess Hill model performance during more general conditions. In particular, we evaluated the ability of the Hill model to describe muscle force responses for both naturally activated and electrically stimulated muscle. In both cases, random length changes with a functionally relevant frequency content were used to provide a more general assessment of model performance.

II. METHODOLOGY

Data were collected from 4 animals. All procedures performed were approved by the Animal Care Committee at Northwestern University.

A. Surgical preparation

Initial surgical preparations were done under deep gaseous anesthesia, according to standard procedures in our lab [6]. In the left hindlimb, the nerve to the soleus was carefully isolated and left in continuity. All other nerves in the distal hindlimb were cut, as were the nerves to the semitendinosus, semimembranosus, and biceps femoris. The soleus tendon was attached to a computer-controlled muscle puller via a bone chip from the calcaneous. Ipsilateral dorsal roots from L4 to S2 were transected to eliminate sensory feedback from the soleus muscle. Contralateral dorsal roots were left intact, as were all ventral roots. After a precollicular decerebration was performed, the gaseous anesthesia was discontinued and the animal was allowed to breathe room air. At the end of the experiment, the animals were sacrificed with a lethal dose (100 mg/kg i.v.) of pentobarbital.

B. Muscle Activation

Both electrical stimulation and natural activation were used to control soleus muscle force. Electrical stimulation was applied using fine stainless steel wires in the proximal and distal portions of the muscle belly. Stimulus trains with constant interpulse intervals (IPIs) and random IPIs were tested at average rates of 10, 20, 30, and 100Hz. Natural soleus activation was obtained by exciting the crossed-extension-reflex (CXR) from the contralateral leg using manual skin compression at the ankle and knee joints to evoke a steady noxious stimulus.

C. Protocols

These experiments were designed to evaluate the Hill model's ability to predict muscle force responses during movement. Random muscle length changes were used to obtain a broad measure of the model's capabilities. Most trials consisted of length perturbations with a bandwidth of 5Hz and amplitudes of ±1mm or ±8mm, centered about an operating point 8mm less than physiological maximum. In a single cat, bandwidths of 2.5Hz and 10Hz and an amplitude of ±4mm were also tested to differentiate between the effects of changes in muscle length and muscle velocity. For each perturbation, force responses were measured at average stimulation rates of 10, 20, and 30Hz. During the CXR trials, a range of muscle forces was obtained by varying the level of the noxious stimulus. The muscle puller was instrumented to measure muscle length and force in all trials.

D. Selection of Crossed-Extension trials

A potential problem with CXR activation is that it might vary during the course of measurement. Because it is difficult to detect activation changes during movement visually, we

0-7803-7211-5/01$10.00©2001 IEEE

Report Documentation Page

Report Date	Report Type	Dates Covered (from... to)
25 Oct 2002	N/A	-

Title and Subtitle Hill Muscle Model Performance During Natural Activation and Electrical Stimulation	**Contract Number**
	Grant Number
	Program Element Number
Author(s)	**Project Number**
	Task Number
	Work Unit Number
Performing Organization Name(s) and Address(es) Department of Physiology Northwestern University Medical School Chicago, IL	**Performing Organization Report Number**
Sponsoring/Monitoring Agency Name(s) and Address(es) US Army Research, Development & Standardization Group (UK) PSC 802 Box 15 FPO AE 09499-1500	**Sponsor/Monitor's Acronym(s)**
	Sponsor/Monitor's Report Number(s)

Distribution/Availability Statement
Approved for public release, distribution unlimited

Supplementary Notes
Papers from 23rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society, October 25-28, 2001, held in Istanbul, Turkey. See also ADM001351 for entire conference on cd-rom.

Abstract

Subject Terms

Report Classification unclassified	**Classification of this page** unclassified
Classification of Abstract unclassified	**Limitation of Abstract** UU

Number of Pages
4

developed an automated procedure to screen out trials where the CXR activation was likely to have changed. This was done by comparing the force responses during natural activation to those obtained with electrical stimulation. Because the cat soleus consists solely of slow fibers, we assumed that the force response during natural activation could be approximated by a combination of the responses to constant electrical stimulation, with rates spanning the range of physiologically relevant firing frequencies. An optimization algorithm was used to minimize the squared error between the CXR responses and the constant electrical stimulation responses at rates of 10, 20, and 30Hz. Only trials with averages RMS errors less than 20% were selected for further processing. Using these limits, approximately 30% of the collected CXR trials were kept for subsequent processing. Changing the acceptable errors limits to 10% or 30% did not affect the qualitative results of this study, but did influence the number of trials available for analysis.

E. Hill Model Estimation

The simplest possible Hill-type model, model consisting of a contractile component in series with an elastic element, was used [6, 7]. The contractile element of this model produces force according to Equation 1, where $A(t)$ is the muscle activation, $F_{LT}(L)$ is the muscle length-tension relationship, and $F_{FV}(V)$ is the muscle force-velocity relationship. Based upon previous results [6], the series elastic element was modeled as a piecewise exponential spring.

$$F_{CE} = A(t) \cdot F_{LT}(L) \cdot F_{FV}(V) \tag{1}$$

The Hill model parameters describing $F_{LT}(L)$, $F_{FV}(V)$ and the series elasticity were measured directly for each muscle, as described in detail previously [6]. A total of 9 parameters were used to describe these model components. In contrast, activation, $A(t)$, is difficult to define in Hill-type models. To avoid an arbitrary model of the activation process while still generating a physiologically realistic activation pattern, activation was defined using the experimental data. To achieve this, each stimulation pattern was applied in the isometric state and force was measured. Activation was then defined as the input required to cause the Hill model to exactly recreate this force using the previously established parameters. Because the equations describing the Hill model are one to one functions they can be inverted to solve for $A(t)$. Timing between isometric and movement trials was carefully controlled to avoid fatigue and potentiation.

The equations describing the Hill-type model were numerically integrated using a fourth order Runge-Kutta method. The inputs to the model were the muscle length, muscle velocity and $A(t)$. Errors between experimentally measured muscle force and that predicted by the Hill model were quantified in terms of percent root mean square (RMS) values, as shown in Equation 2.

$$Error = \sqrt{\frac{\sum_N \left(F_{\exp t} - F_{Hill}\right)^2}{\sum_N F_{\exp t}{}^2}} * 100\% \tag{2}$$

III. RESULTS

Figure 1 summarizes the results during constant stimulation. Part A shows typical data for one cat with 10, 20, and 30 Hz stimulation. Results on the left are for ±1mm displacements while those on the right are for ±8mm displacements. The dots below each force trace indicate stimulation times. Thin lines represent measured muscle forces and thick lines represent forces predicted by the Hill model. Differences between these curves are the Hill model errors. Part B summarizes these errors for the three animals in which these experiments were performed. Percent RMS errors are plotted as a function of stimulation frequency for both perturbation amplitudes. Circles indicate actual data points. Open circles correspond to a different random length pattern (identical bandwidth and amplitude characteristics) tested in a single cat. Hill model errors were largest at low stimulation frequencies and larger displacement amplitudes. Random IPIs and different length randomizations did not alter these conclusions.

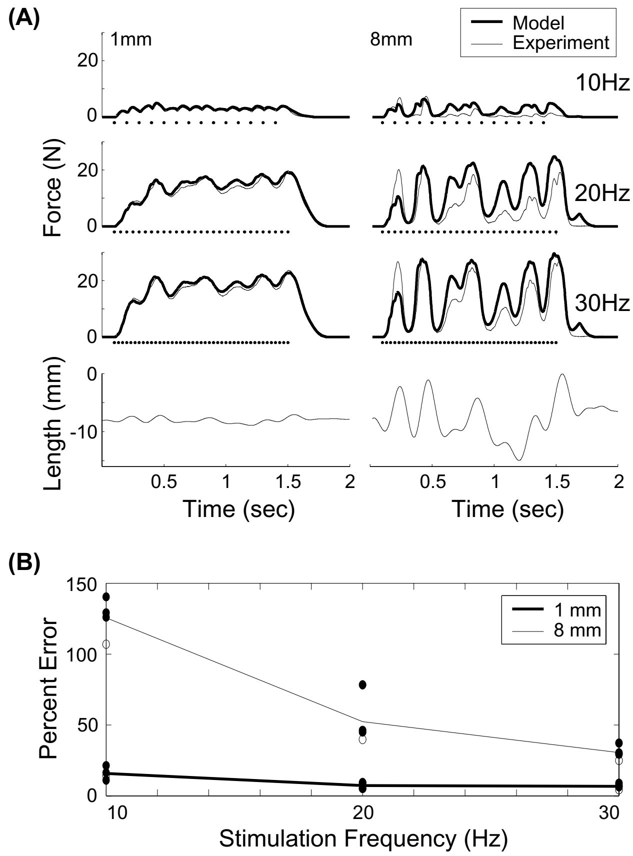

Figure 1. Hill model errors for constant electrical stimulation.

The decreased Hill model performance with increased muscle excursion could be due to either amplitude or velocity effects. To determine which of these factors was most important, a more complete set of perturbation bandwidths was tested in one animal. Figure 2A summarizes the results of these experiments. Percent RMS errors for each combination of perturbation bandwidth and amplitude are plotted as a function of stimulation frequency. Perturbation bandwidth was varied by changing the output rate of the displacement sequence. Therefore, muscle velocity was proportional to the product of the bandwidth and amplitude.

Figure 2A shows that trials with identical muscle velocities had nearly identical Hill model errors, even though displacement amplitudes differed. Note for example the errors associated with the 5Hz, 8mm trial and the 10Hz, 4mm trial. These results indicate that increased Hill model errors are associated primarily with increases in muscle velocity. Figure 2B shows the model errors as a function of RMS muscle velocity for each of the tested stimulation rates. The results indicate that, at each stimulation frequency, there was a nearly linear relationship between muscle velocity and Hill model performance.

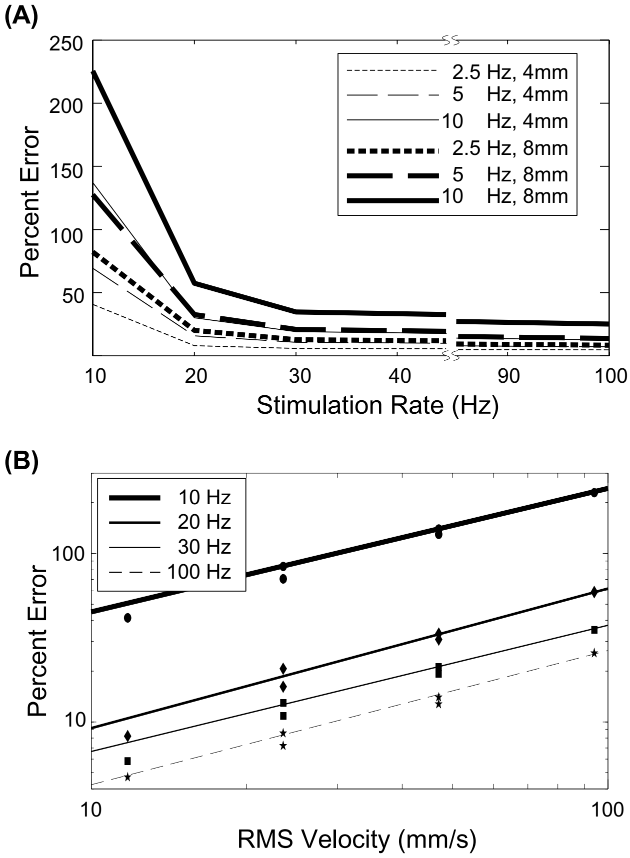

Figure 2. Velocity related errors during electrical stimulation.

Figure 3 shows the results when the soleus muscle was activated naturally via the crossed-extension reflex. Figure 3A shows a typical force response to an applied perturbation (thick line) and the corresponding Hill model prediction with a constant activation input chosen to match the CXR force level before perturbation onset. Steadiness of activation during these trials was assessed by matching the CXR response to an optimized combination of responses with constant electrical stimulation, as described in the Methods. The medium weight line shows the optimal combination of stimulation responses matched to this trial. The close match indicates that the muscle activation via the crossed-extension reflex was nearly constant during the course of the applied perturbation, indicating that variations in muscle activation did not contribute significantly to Hill model errors. Figure 3B summarizes the Hill model errors for this cat as a function of the pre-perturbation force level. Percent RMS errors were calculated over the course of the imposed movement, and decreased with increasing force level. At low force levels, the magnitudes of these errors were slightly higher than those obtained with 10Hz constant stimulation. Errors decreased with increasing stimulation level. Most errors fell between those measured with 10 and 20 Hz constant stimulation. A possible explanation for these large error magnitudes is that the motor units in naturally activated muscle fire predominantly in the 10-20Hz range. The actual firing rate distributions can be estimated using the optimization results used to select trials with steady activation by examining the weights chosen for each stimulation frequency (See Methods). Figure 3C illustrates how the estimated motor unit firing rates varied as a function of muscle force for this animal. The force contributed by motor units firing at approximately 10, 20, and 30Hz is plotted as a function of total muscle force. Over the forces tested, which went up to 80% of the maximum tetanic force, most force was contributed by motor units firing between approximately 10-20Hz. As would be expected, the higher firing rates contribute more at higher force levels.

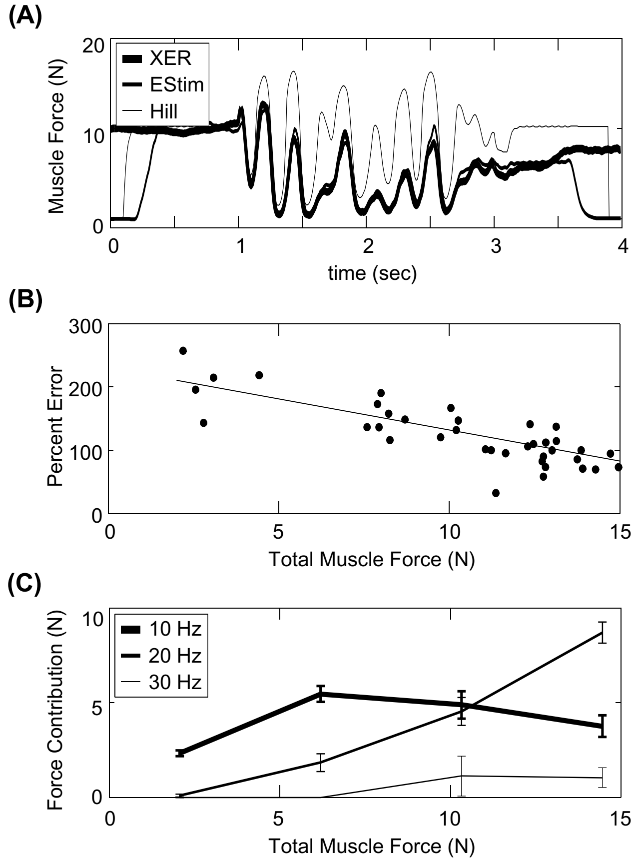

Figure 3 Crossed-extension results

IV. Discussion

This work evaluated the accuracy of the Hill model during functionally relevant conditions, including muscle activation via low frequency electrical stimulation and the crossed-extension reflex, and a range of physiologically relevant random length changes. Our results indicate that Hill model errors increase dramatically with increases in muscle velocity and decreases in stimulation frequency. The errors during natural activation were large and most closely resembled those obtained with constant stimulation

frequencies of 10-20Hz. For large muscle excursions, the errors with naturally activated muscle typically exceeded 50%, indicating that the use of the Hill model is not appropriate for these conditions. These results were robust with respect to different length randomizations and stimulation patterns with variable IPIs, suggesting that our conclusions are general in nature.

Hill model errors were highly dependent upon muscle velocity (see Figure 2). Increases in muscle velocity resulted in nearly linear errors increases. Given these results, it is possible to estimate Hill model performance for a range of movement conditions provided that estimates of muscle velocity and motor unit firing rates can be obtained. The relationship between muscle velocity and Hill model errors also suggests that model performance could be drastically improved by incorporating additional velocity effects. Similar conclusions have been reached by other researchers. Shue and Crago [4, 5] incorporated nonlinear coupling between muscle velocity and activation to improve Hill model performance for electrically stimulated cat soleus. Brown and Loeb [8] reached similar conclusions for the cat caudofemoralis muscle. Future work will examine if these improved models provide significant improvements in muscle force prediction under the general test conditions used in this study.

This is the first study to assess Hill model performance for naturally activated muscle. One of the difficulties with using natural activation is determining whether or not activation varies during the course of the applied perturbation. To circumvent this problem, we used an optimization routine that compared CXR trials to a combination of electrically stimulated trials. This technique was able to match CXR trials well (see Figure 3A). In addition to providing a quantitative method for trial selection, this algorithm provided estimates of the motor unit firing rate distributions within the muscle. These were predominantly between 10-20Hz for the range of forces tested. This range of firing frequencies corresponds to where the largest Hill model errors were obtained with electrical stimulation, thereby explaining the large errors observed when comparing for naturally activated muscle force to that predicted by the Hill model.

V. Conclusions

Together, the results of this study place bounds on the accuracy that can be expected from Hill model of muscle force generation for both electrically stimulated and naturally activated muscle. In general, large model errors can be expected except for small movements and near-tetanic contractions. In other instances, improved muscle models should be used.

VI. Acknowledgements

This material is based upon work supported by NIH grant 5R01AR41531-07 to CJH and TGS, NIH Training Grant 5 T32 HD07418, and a National Science Foundation grant awarded in 2001 to EJP.

VII. References

[1] J. M. Winters, "Hill-based muscle models: a systems engineering perspective," in *Multiple Muscle Systems: Biomechanics and Movement Organization*, J. M. WInters and S. L. Woo, Eds. New York: Springer-Verlag, 1990, pp. 69-93.

[2] F. E. Zajac, "Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control," *Crit Rev Biomed Eng*, vol. 17, pp. 359-411, 1989.

[3] Jewel and Wilkie, "The mechanical properties of relaxing muscle," *J Physiol (Lond)*, vol. 152, pp. 30-47, 1960.

[4] I. E. Brown and G. E. Loeb, "Measured and modeled properties of mammalian skeletal muscle: IV. dynamics of activation and deactivation," *J Muscle Res Cell Motil*, vol. 21, pp. 33-47, 2000.

[5] G. H. Shue and P. E. Crago, "Muscle-tendon model with length history-dependent activation-velocity coupling," *Ann Biomed Eng*, vol. 26, pp. 369-80, 1998.

[6] T. G. Sandercock and C. J. Heckman, "Force from cat soleus muscle during imposed locomotor-like movements: experimental data versus Hill-type model predictions," *J Neurophysiol*, vol. 77, pp. 1538-52, 1997.

[7] A. M. Krylow and T. G. Sandercock, "Dynamic force responses of muscle involving eccentric contraction," *J Biomech*, vol. 30, pp. 27-33, 1997.

[8] I. E. Brown, E. J. Cheng, and G. E. Loeb, "Measured and modeled properties of mammalian skeletal muscle. II. The effects of stimulus frequency on force-length and force-velocity relationships," *J Muscle Res Cell Motil*, vol. 20, pp. 627-43, 1999.